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Thermo-Controlled Device for Inducing Deep Coagulation in the Liver With the Nd:YAG Laser

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Background and Objective: **To increase the effectiveness of laser-induced interstitial thermotherapy (LITT), a new thermo-controlled application system for minimal invasive intervention was designed. Our system consists of a laser applicator of 2.5 mm in diameter, insertion equipment, and a Nd:YAG-laser source.**
Study Design/Materials and Methods: **A cylindrical light emitting fiber (1–6 cm in length) was placed in the center of the applicator. The surrounding tissue was irradiated through a Duran® window at the distal end of the applicator. The power of the laser source was controlled dynamically by thermosensors in a water-cooling system of the laser applicator. The temperature at the surface of the Duran® window was kept constant at ~60°C, without charring the surrounding tissue.**
Results/Conclusion: **We obtained homogeneous coagulation zones. In in vitro experiments with pig livers, we reached ellipsoid coagulation volumes of 3 and 5 cm in diameter within 10 minutes, corresponding to a volume of ~25 cm^3. Lasers Surg. Med. 20:149–156, 1997.** **© 1997 Wiley-Liss, Inc.**

Key words: Nd:YAG-laser surgery; laser-induced interstitial thermotherapy

INTRODUCTION

The thermal effect in biological tissue is a fundamental property of infrared laser radiation. Superficial areas easily can be photocoagulated by contact or noncontact light application. Deep tumors or metastases can be treated only by surgery, drugs, or locally applied toxic substances.

In inoperable localized tumors, laser-induced interstitial thermotherapy (LITT) is an effective and investigative method of tumor therapy. Different types of tumors in the liver, pancreas, lung, and brain have been treated in the past [1–6]. Benign hyperplasia (prostate) also can be reduced by laser coagulation [7–9]. Coagulative and hyperthermic effects due to photon absorption and heat conduction lead to an immediate or delayed tissue destruction [5].

In the past the Neodymium-doped Yttrium Aluminium Garnet (Nd:YAG, 1064 nm) laser mainly served as a light source, because of its deep light penetration in biological tissue [10,11]. The light application system is equally important to reach an effective coagulation zone within a few minutes. In the beginning of LITT, bare fibers were used to induce interstitial hyperthermia [12]. The high power density at the distal end of the fiber restricted the applicable laser power to a few watts only in order to avoid carbonization. In result, only small coagulation zones with diameters of up to 1.5 cm could be achieved [13,14]. To increase the coagulation volume, several (modified) bare fibers were spread out over the tumor volume [6]. The development of radially or diffusely emitting fibers enlarged the effected zone to diameters of up to 1–2 cm, but there was still

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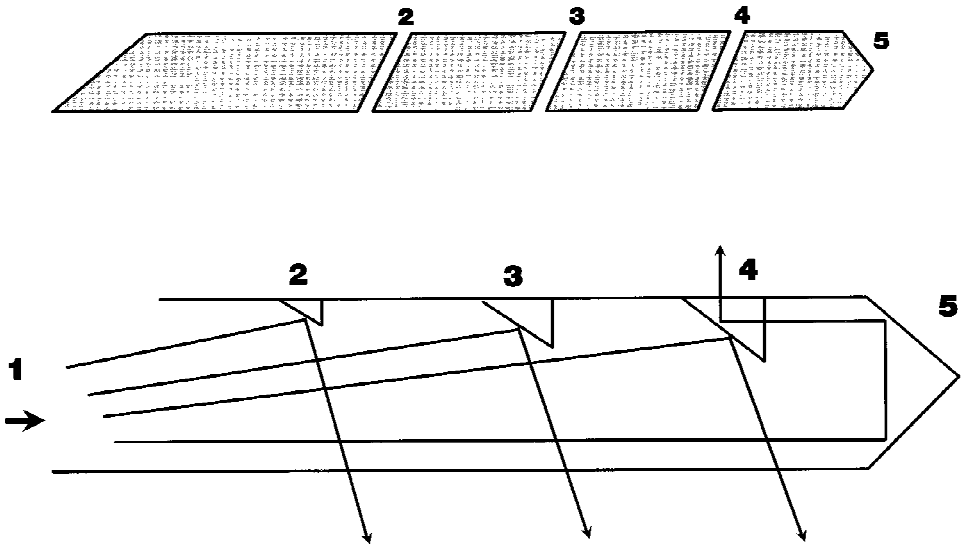

Fig. 1. Scheme of a modified distal fiber end. Circular grooves [2–4] are of increasing depth reflecting more and more radiation from the axis path [1] to the radial direction. A conus [5] at the distal end of the fiber reflects remnant rays backward.

the limitation of laser power to a few watts [15–18]. Open liquid cooling of the fiber tip has been used, resulting in a liquid pool with increased interstitial pressure [19,20]. Others investigated a frosted sapphire tip to obtain a wider angle of illumination [16,22]. However, the width of the sapphire tip and the compound metal collar limited these tips to percutaneous applications. The initial use of coaxial gas flow to cool the fiber tip is now obsolete, because of air embolism [21]. Recently, closed water-cooled laser devices were developed in order to increase the laser power [22,23]. These devices do not provide the possibility of monitoring the real time temperature of the surrounding tissue; furthermore the amount of applied power is still limited.

To increase the effected volume and to reduce therapy time, further investigations and a new applicator design were necessary.

MATERIALS AND METHODS

Tissue and Laser Light

We used the livers of killed pigs for all in vitro experiments. To study the effect of water cooling in tissue, we applied Nd:YAG-laser light (1064 nm) via a fiber to the liver in varying degrees of intensity. Ice water (1–4°C) was used to cool the surface. The water was pumped via a roller pump to the laser light contact area of the liver. The coagulated volumes were calculated by three-dimensional measurements. In a second series of experiments we used our thermo-controlled application system for liver coagulation. The temperature in the tissue was measured by thermoelements at a distance of 5 mm and 10 mm from the applicator surface. Again, the coagulation volume was measured three-dimensionally.

Fiber for LITT

The distal end of a 600-μm fiber was modified by adding circular grooves. The grooves at the distal end are of increasing depth, as shown in Figure 1. If the inclination angle of the frontal area of the grooves with respect to the fiber axis is larger than the angle of total reflection between the fiber core and air, nearly all rays coming through the fiber from the proximal end will be totally reflected and thus exit radially. Increasing the depth of each groove toward the distal end, more and more radiation is reflected from the axis path to the radial direction. At the distal end of the fiber, a cone reflects the remnant rays backward, leaving the fiber at one of the grooves radially. This creates a defined cylindrical radiated volume that is much larger than that of a point source.

We modified fiber tips from 0.5–6 cm. In our experiments we used a 4 cm modified fiber tip.

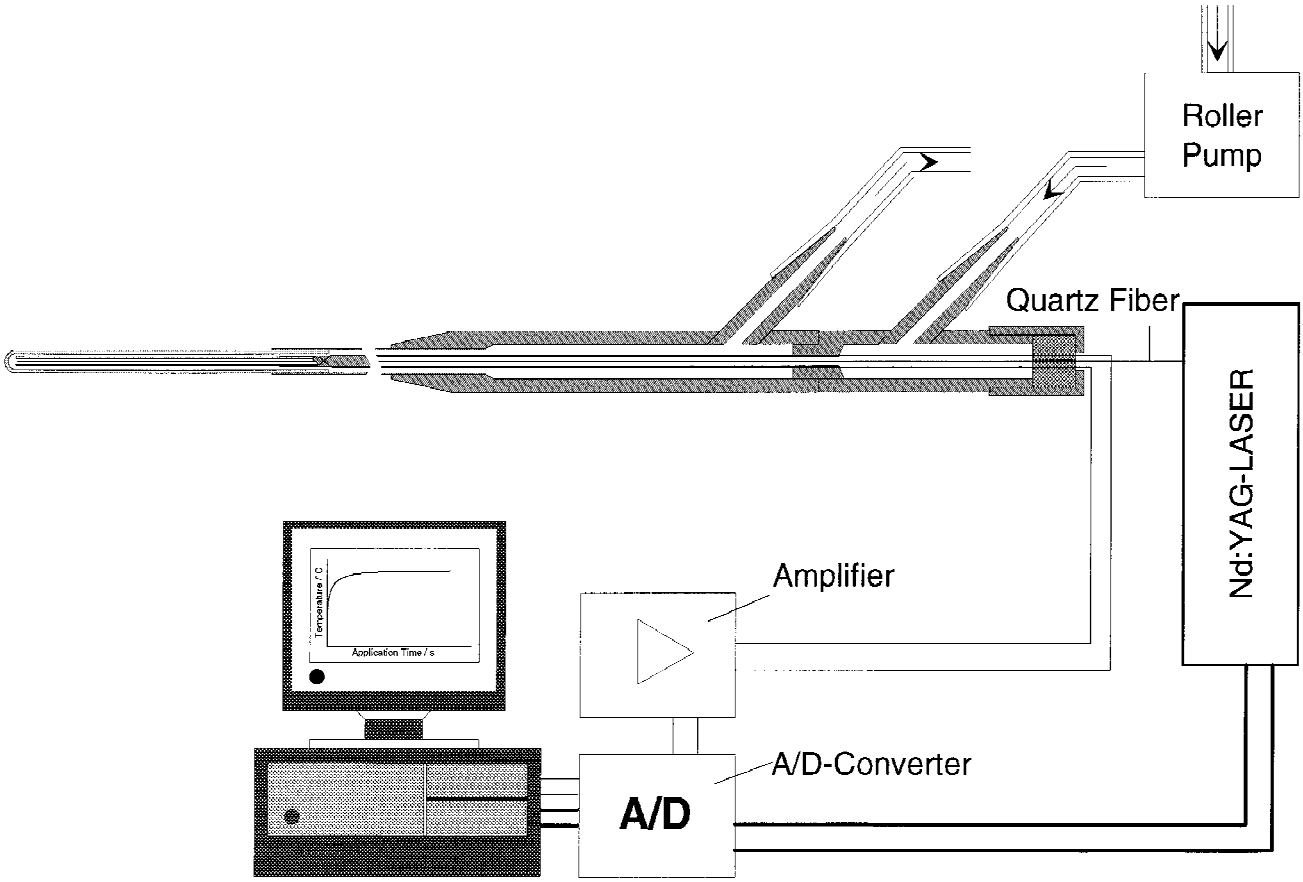

Fig. 2. Laser application system, which consists of a Nd:YAG-laser, a laser device, and a computer controling unit.

LITT Application System

The laser device has been designed to deliver high laser power to deep tissue areas as well as to protect the optical fiber. The whole system is shown in Figure 2. A handpiece comprises an axial channel for fiber deposition and two coaxial ducts for the rinsing circulation. The large distal part of the applicator (20 cm in length and 2.5 mm in diameter) ends with a quartz window. The modified laser fiber tip is hermetically sealed against its surroundings by an optically transparent case. The case contains one optical transparent tube and one less absorptive tube, which are the inlet and outlet channels for the rinsing fluid.

The rinsing medium flows along the inner surface of the case and is heated by the irradiated tissue. The flow rate is kept constant by a roller pump (30 ml/min). The characteristic temperature rise is measured by an thermo-element fixed in the outflow channel at a spot where the heating by direct radiation is negligible. The true temperature of the tissue in the border area can be computed from the temperature rise in the outflowing medium. This is done to control the laser power automatically and to modulate it continuously in order not to exceed the critical temperature and to avoid drying out and charring. The laser power is controlled by an electronic thermomonitoring supplement to the laser. The process is not fixed or predetermined. It is performed dynamically with software support, triggered by the operator via a laser pedal. The software computes the actual laser power taking into account the temperature course and the critical tissue temperature.

RESULTS

When laser light was applied to the surface of the water-cooled liver, the coagulation zone was found to be deep inside the liver tissue. We found the enlargement of the coagulation volume to be directly dependent on the laser energy and application time. The uncoagulated zone at the

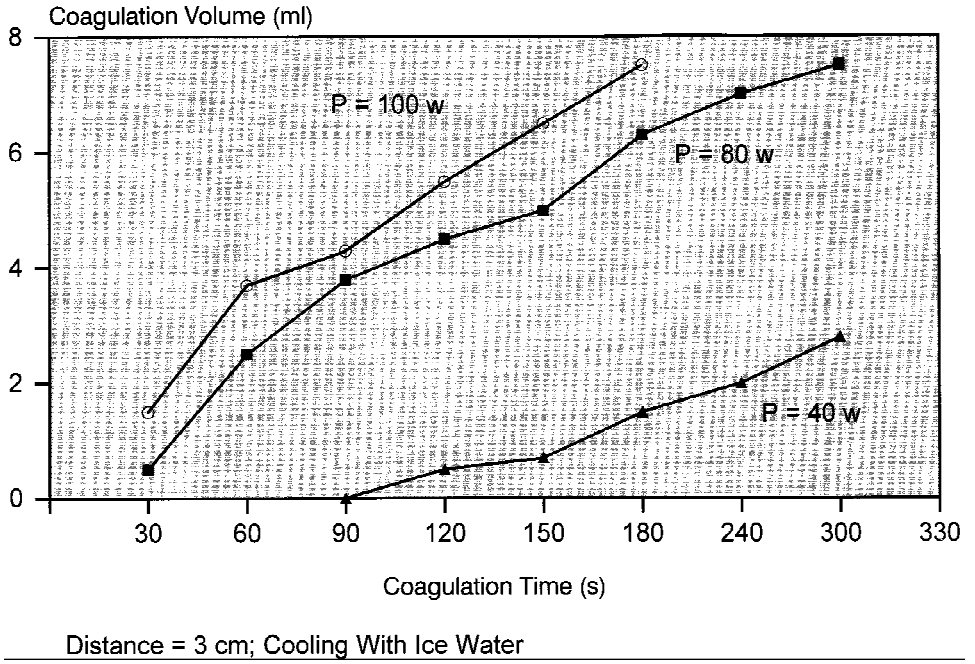

Fig. 3. Relationship among laser power, time, and coagulation volume when the liver surface was cooled with ice water. The distance between fiber and surface was 3 cm.

tissue surface could be preserved when the laser power density was low. In fact, there was a correlation between laser power and cooling effect. Heat conduction and heat flow could be maintained when there was a steady correlation between these two effects. When the laser power increased the uncoagulated zone at the surface disappeared (Fig. 3). In result, the coagulation volume was much bigger than without cooling.

We also found a correlation between the initial laser power and the maximum coagulation volume. High laser power at the beginning resulted in a bigger coagulation volume when the cooling effect was constant (Fig. 4). The coagulation volume was limited when the temperature inside the coagulation zone rose. Water vaporization ($\geq$100°C) resulted in an explosion of the liver coagulation zone. When the applied laser power was reduced by thermo-controlling (<100°C), the coagulation volume further increased. On the basis of these results, a new device for interstitial laser light application was constructed. We combined the water cooling effect with low laser light density, using a cylindrical light emitting fiber. To protect the tissue from carbonization and water vaporisation, we used thermo-sensors in the backflow of our water rinsing system and in the tissue at a distance of 5 and 10 mm from the applicator. We found a direct correlation between the temperature at the applicator surface and the water temperature in the backflow (Fig. 5). However, the temperature at the applicator surface corresponds directly with the tissue temperature in this area. There is minimal absorption of laser energy in water when Nd:YAG-laser (1064 nm) light is used. Consequently the rinsing temperature in the backflow indicates the temperature of the surrounding tissue. In our experiments we found that 28°C in the water backflow corresponded to 60°C in the surrounding tissue. This temperature was chosen to avoid tumor spread caused by water vaporization.

In further experiments we kept the temperature in the backflow of the water rinsing system constant at 28°C by dynamic modulation of the laser power applied. We started with 70 watts reducing laser power step-by-step in correlation to the water temperature. After 5 minutes, only 25 watts were needed for a continuous heat flow (Fig. 6). With this system we reached a increasing homogeneously coagulation volume. Within 10 minutes, we were able to produce a coagulation zone of 3 cm in diameter and 5 cm in length without any central carbonization (Fig. 7).

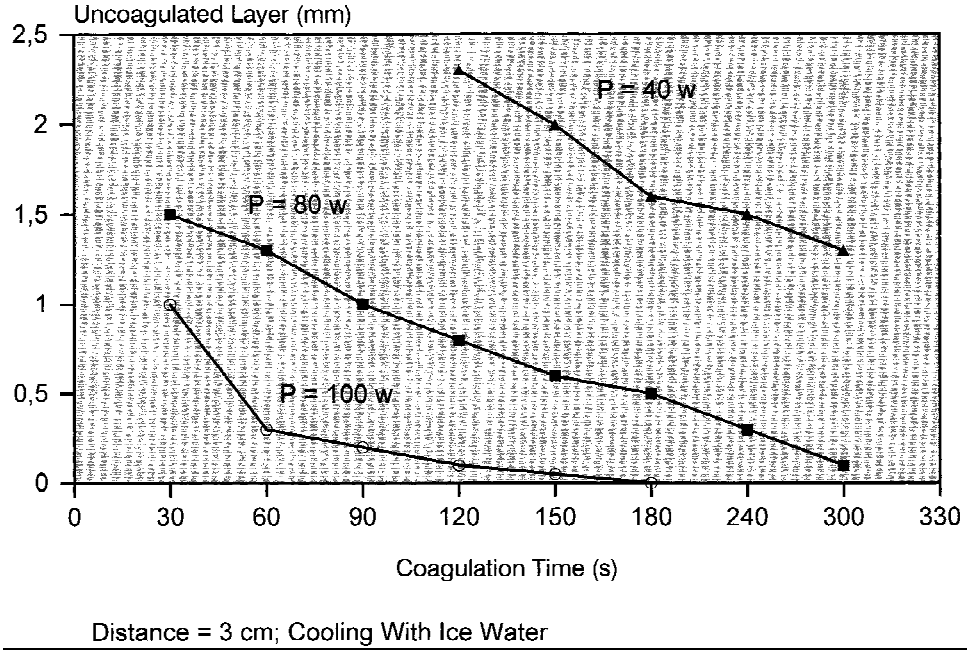

Fig. 4. Relationship among laser power, uncoagulated tissue layer, and time when the liver surface was cooled with ice water. Distance between fiber and surface was 3 cm.

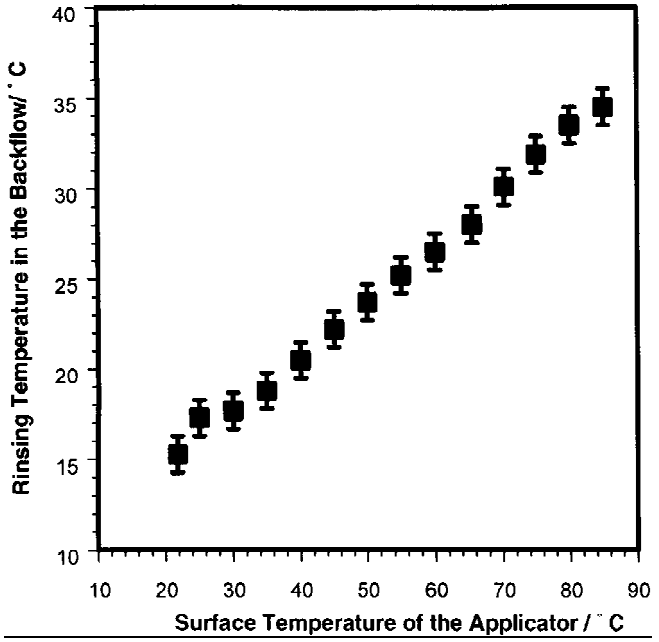

Fig. 5. Correlation between applicator surface temperature and the water temperature in the backflow of the rinsing system.

DISCUSSION

The critical zone for laser light application in biological tissue is the zone of light entrance. The correlation between power density, quality of light, and tissue properties results in a rise in temperature. Carbonization of the coagulation zone further increases the temperature up to vaporisation. The properties of carbonized tissue layers change, absorbing more and more energy. This promotes charring and limits the further growth of the coagulation zone by direct laser heating.

In our research efforts we studied the positive effect of surface cooling during laser light application. We found the coagulation zone and peak temperature shifted into the depth of the tissue without any damage to the surface. This effect might be useful in the treatment of benign hyperplasia, as, e.g., in the prostate. With an effective cooling system, the urethra can be spared while deeper layers are coagulated.

In our tumor model, further radiation resulted in a large coagulation volume when the total energy was applied. However, we had to reduce laser power whenever the temperature in the coagulation zone went up. Not only water vaporisation with explosion of the organ, but also carbonisation of surrounding tissue prevented further maximization of the coagulation volume. Feedback via thermo-sensors in the coagulation zone helped to maximize the coagulation zone

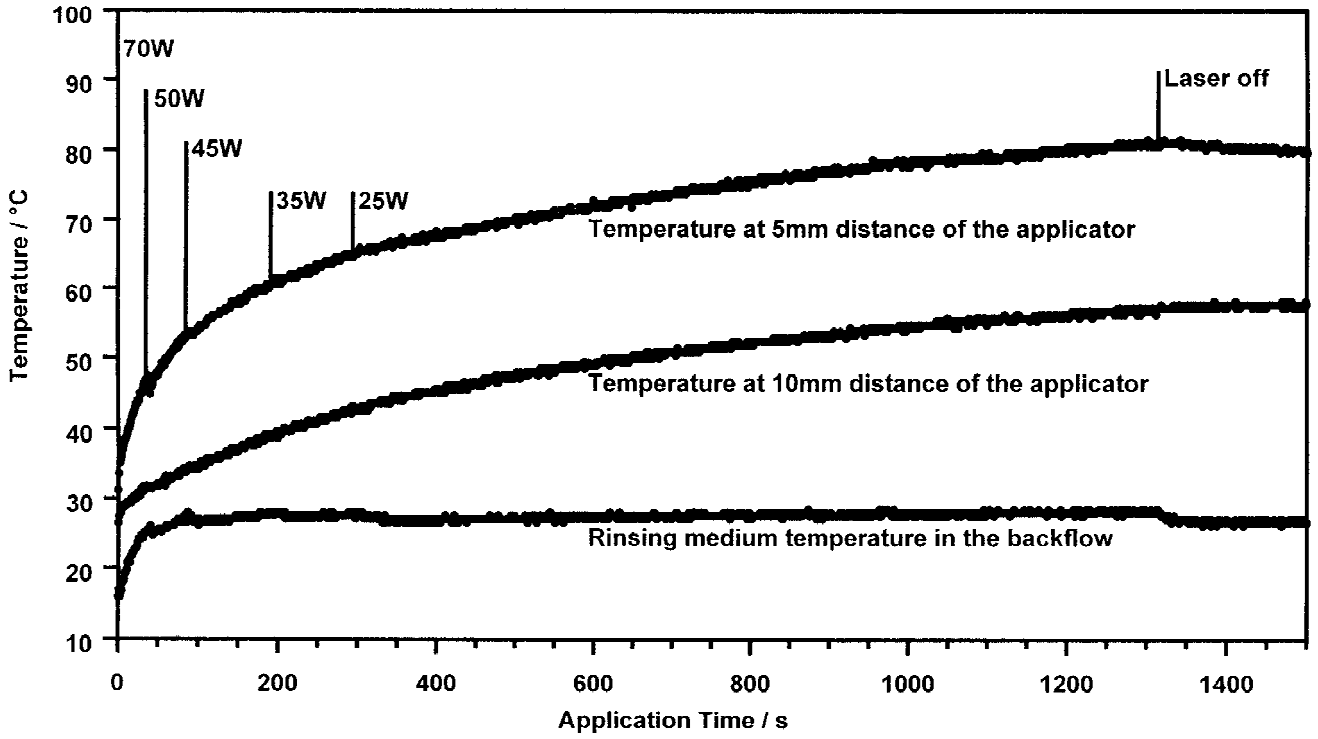

Fig. 6. LITT temperature during in vitro coagulation. Relationship between the temperature in the backflow of the water rinsing system and the coagulation zone at 5 and 10 mm distance from the device. LITT was performed dynamically starting with 70 watts.

when the laser power was reduced and the temperature kept constant at <100°C.

In this way a new laser device was designed for local interstitial laser thermotherapy. The length of the handpiece allows access to deeper parts of the liver, and the small diameter minimizes the damage to the normal tissue. The cooling system protects the fiber and prevents damage to fiber tip. This positive effect is also described by other authors [22,23]. However, coagulation volumes remained small (10–20 mm in diameter) when a point source was used. To provide contact with the surrounding tissue, we modified the distal end of the fiber and obtained a cylindrical light-emitting source. The laser power density was much lower, allowing high initial laser energy. Roggan [24] used a similar "zebrafibre" for cylindrical light emission with a similar effect. Different lengths of the modified distal fiber tips (0–6 cm) can be used for different types of tumors.

The correlation between the rise in temperature measured from the backflow of the water rinsing system and the surface temperature of the applicator was directly proportional. This direct information about the temperature at the applicator window was used for automatic regulation of the laser power. The initially high laser power corresponds to deep tissue coagulation. The warming up of the surrounding tissue corresponds to the absorption of the laser irradiation and the heat conduction from the warmer parts of the coagulation zone. Reducing laser power dynamically, corresponding to the critical tissue temperature, we obtained a steady state between energy supply and energy needed for the coagulation process. Carbonization in the surroundings of the applicator could be avoided by keeping the tissue temperature constant below 100°C. In result, we obtained a continuous heat flow and a homogeneous growth of the coagulation volume.

This laser system can be used for coagulation of unresectable liver metastases. The small diameter of the applicator allows percutaneous as well as laparoscopic use. Also, brain tumors can be punctured and laser-treated, simultaneously controlled via magnetic resonance tomography. The integrated cooling system and the cylindrical or halfcylindrical light emmision predispose the laser system for the treatment of benign hyperpla-

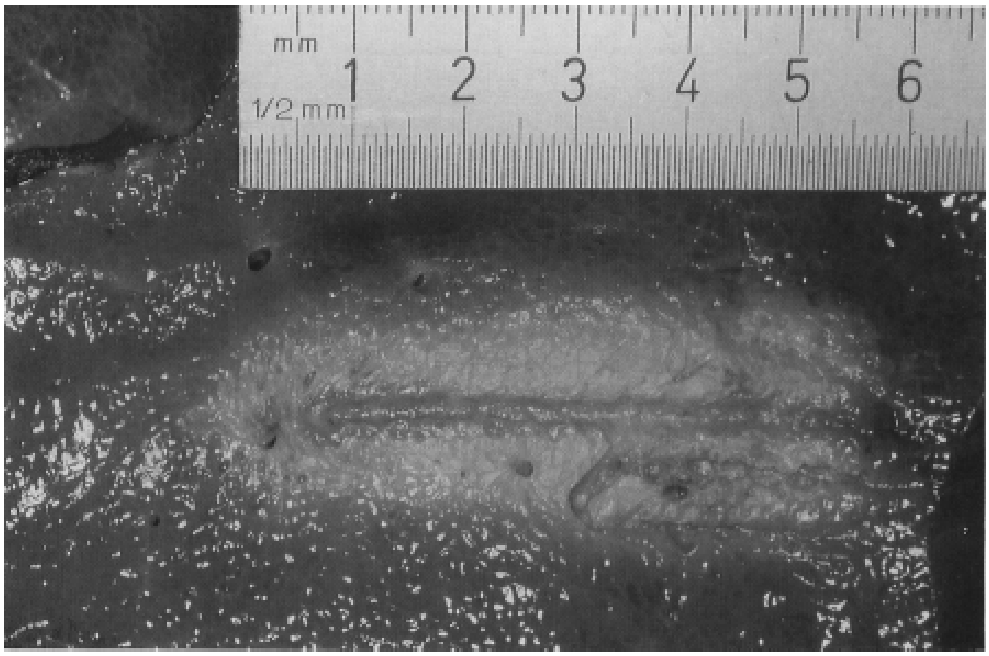

Fig. 7. Liver coagulation zone without carbonization 10 minutes after dynamic LITT application.

sia of the prostate. Further investigations will show the effectiveness of this applicator-system in in vivo experiments.

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